Introduction

Force instrumented treadmills facilitate online kinetic measurement of a high volume of steps in a small space with the safety of support harnesses (Merholz and Elsner, 2014) and, combined with visual projection, can allow practice of altering walking in response to cues (e.g. stepping to targets, over or around obstacles (Heeren et al., 2013). For these reasons use of instrumented treadmills for rehabilitation and clinical assessment is increasing (Bank et al., 2011; Duysens et al., 2012; Heeren et al., 2013; Hollands et al., 2014; Hollands et al., 2013; Mazaheri et al., 2015; Mazaheri et al., 2014; Peper et al., 2015; Timmermans et al., 2016; van Ooijen et al., 2015; Weerdesteyn et al., 2006).

Single uniaxial force instrumentation of the treadmill belt affords centre of pressure (CoP) gait event detection (GED) as a proxy for gold standard kinetic (dual, multi-axial, force-plates) or kinematic GED. CoP GED has been shown to correspond well with kinematic GED during steady-state treadmill walking in young healthy adults (Roerdink et al., 2008). However, it is not known whether CoP GED corresponds with kinematic GED when steps are altered in response to environmental cues, or when alterations in CoP trajectories occur due to pathology (i.e. stroke (Wong et al 2004)).

To support valid gait assessment in the context of growing treadmill use in clinical assessment, this study aimed to determine if there are differences in CoP and kinematic GED in young healthy (YH) and stroke survivors (SS) during treadmill walking. We compare GED methods in the walking condition of varying steps; the context in which they are increasingly being applied. Specific questions are:

1) Are there significant differences between methods within groups?

2) Are differences between methods greater in SS than YH (and according to paretic and non-paretic limbs)?

Methods

Participants

YH, aged 18-35 years, were recruited by poster advertisement across the University. SS were recruited from community stroke support and exercise groups in Greater Manchester. Participants were included if they could walk ten-metres within 30s, had no visual impairments preventing sight of stepping targets, and no co-morbidities affecting walking.

The University of Salford, College of Health and Social Care Research Ethics Committee approved the study, and all participants provided written informed consent.

Procedures

Self-selected walking speed (SSWS), functional mobility (10m walking test (Green et al., 2002); Timed Up and Go (Hiengkaew et al., 2012) and Dynamic Gait Index (Jonsdottir and Cattaneo, 2007)) were collected to ascertain mobility status of the SS.

Participants were acclimatised with walking on the treadmill without stepping targets for approximately 3 minutes. Each participant's SSWS was determined by increasing speed from 1km/h until participants were walking faster than preferred, then decreasing speed to a comfortable pace. Participants walked to targets located at their usual step lengths and widths (established when walking during earlier no-target acclimatisation period) for 1 minute, to become acquainted with target stepping. Step characteristics such as speed, step length and width were recorded as a basis for programming the location of targets for subsequent personalised target-stepping tasks.

Participants stepped to targets located according to their personalised protocol, projected on the treadmill belt while walking at SSWS (figure 1) according to a previously described paradigm (Hollands et al., 2015). 12 targets (8cm wide x40cm long) were projected at preferred step length and 12 of the same size for both shortening and lengthening steps (±25% of preferred step length). A further 24 targets of different shape (20cm wide x15cm long) were projected on the midline of the treadmill to elicit narrow foot placements. Participants were not allowed to use a handrail for stability; however, SS wore a harness for safety.

Kinetics

Signals from a single large (0.8x3.0m) uniaxial force plate was conditioned (100Hz low-pass filter) and recorded at 500Hz using CueFors1 software in the C-Mill (MotekforceLink, Culemborg, The Netherlands). CueFors1 analyses CoP cyclogram, also defined as gaitogram (Roerdink et al., 2014)(Figure 2), to generate gait events.

Kinematics

Kinematics were collected with a six-camera motion capture system (Qualysis, Gothenburg, Sweden) at 126Hz for healthy participants and at a minimum sampling rate of 31Hz for SS (due to synchronisation of high speed video for some participants); kinematic data was subsequently spline interpolated to 500Hz to match the C-Mill data. Toe and heel markers on the 2nd distal phalangeal head and the calcaneus were used for kinematic GED. The C-Mill and motion capture systems were synchronised with an electronic pulse generated by CueFors1 software that triggered the start of motion capture. Kinematic gait events were detected offline after interpolating and filtering (2th order bidirectional 6Hz low pass Butterworth filter).

Two GED algorithms were used to define gait events: the first defined FC as the minima of the vertical displacement of the heel marker (VFC) and FO at the maxima in vertical velocity of the heel marker (VFO)(Pijnappels et al., 2001; Roerdink et al., 2008). The second defined FC as the maximum anterior displacement of the heel marker (AFC) and FO as the instant that the anterior velocity of the toe marker is zero (AFO) when it transitions from posterior to anterior velocity (Zeni et al., 2008).

Statistical analysis

At least 30 gait events, FC and FO, were detected by both kinematic and CoP algorithms per participant per foot. Data comprised 10 normal steps (before the adjustment protocol) and 60 adaptation steps (30 per foot). CoP events were matched to the kinematic events occurring within 200ms, if no such match could be made they were recorded as the proportion of steps that could not be matched (unmatched, see table 2).

To determine if there are significant differences between methods within groups: Differences between matched CoP and kinematic gait event for paretic and non-paretic and left and right side of SS versus YH were compared using a one-sample (two tailed) T-test against a reference value of 0ms (i.e. no difference)(Roerdink et al., 2008).

To determine if differences in methods are greater for SS than YH (and according to limbs): differences between methods were compared in a repeated measures ANOVA, separately for each gait event (FC, FO), with between subjects' factor group and within subjects factors methods (CoP-Vertical, CoP-Anterior) and limb.

Results

A total of 7 YH and 13 SS participated (demographics see table 1). No abnormalities in cyclograms which would have prevented CoP GEDs were found on visual inspection of individual participant data (figure 2).

Foot contact

Detailed timings of gait events are reported in table 2. VFC detected FC significantly earlier than CoP in YH ($p<0.001$) but there were no differences between methods in FC detection for SS (on either paretic and non-paretic side). FC via AFC was detected significantly earlier than CoP in healthy participants ($p<0.001$) and in SS on both paretic and non-paretic sides ($p<0.001$ for both).

A significant interaction effect between limb, GED method and group ($F(18)=4.960$, $p=0.039$) indicates that the difference between COP and AFC GED is smaller on the non-paretic side than the paretic side. Additionally, FC identified in stroke survivors using VFC were matched with CoP detections less often (P 20% and nP 9% was unmatched), than AFC across all participants (YH 3%, SS P 7% and nP 4%).

Foot off

The AFO algorithm worked with similar success in both groups and sides (3% unmatched FO). The VFO was less successful with 7% and 11% unmatched FO in SS and YH subsequently. FO was detected earlier in VFO than CoP in all participants ($p<0.001$). FO was detected earlier in AFO compared with CoP in YH and in SS for both paretic and non-paretic sides ($P<0.001$).

A significant interaction effect between limb, GED method and group ($F(1,18)=9.173$, $p=0.007$) was found indicating the difference between CoP and AFO GED is significantly larger on the non-paretic than paretic limbs.

Step times

Phase durations (e.g. swing and stance), calculated using the times of FC and FO, looked similar, on visual inspection (see figure 3), between FC and FO detected with AFC, AFO kinematic criteria and CoP detected gait events. Conversely, temporal gait parameters using VFC and VFO kinematic criteria yielded a significantly shorter stance and longer swing phase (figure 3); as a result of slightly late FC detection and early FO detection in SS (see Table 2).

Discussion

Traditionally, GED is applied during steady state walking on a treadmill/over-ground (Roerdink et al., 2008; Roerdink et al., 2007). However, owing to the importance of adapting steps in response to environmental cues and the increasing use of instrumented treadmills to train and assess gait in this context, we robustly compared the performance of GED methods during step alterations (longer, shorter, and narrowing) for both YH and SS.

We found that, for SS, detecting FC using VFC and FO using VFO (Pijnappels et al., 2001; Roerdink et al., 2008) kinematic criteria failed too often to be considered reliable. Conversely, AFC and AFO kinematic criteria were more successful (Table 2). Where kinematic event detection was successful, there was agreement between methods to within 100ms (Figure 3). This may provide sufficient resolution for CoP GED in many training and assessment applications beyond steady-state walking. However, the kinematic criteria suggested (Zeni et al., 2008) for use with SS treadmill walking (AFO, AFC) had the largest differences with CoP GED for FC and varied according to limb. These contextual differences between CoP and kinematic GED methods are considered further to inform application.

The FC detection based on VFC is determined as the local vertical minimum of the heel. Whilst there is a systematic difference between this and CoP methods, it is small (a difference of 29ms in YH, P 1ms and nP 5ms in SS) and of little practical significance. More importantly, however, many SS have abnormalities in foot and ankle movement (Burridge and McLellan, 2000; Stein et al., 2010) which often result in a mid-FC, with the heel continuing to lower after contact. This would lead to the minimum heel height occurring after FC, as observed here. Such abnormalities may also explain why the VFC method appears to fail more frequently in paretic FC in SS (20%).

The AFC method defines FC as the most anterior position of the heel. In most people, however, knee flexion commences before initial contact (Winter, 1992) resulting in the heel moving posteriorly at initial contact, leading to early detection compared to CoP. This corresponds with our data and suggesting CoP more accurately reflects weight transfer at FC than actual movement (AFC).

VFO identifies the maximum vertical velocity of the heel marker as FO. This is considerably earlier than toe-off and leads to an early (positive) FO detection (Table 2). FO detection using AFO is based on the zero-crossing of the forward velocity of the toe. Because the toe marker in our model is placed on the base of the 2^{nd} metatarsal on the shoe, the shoe could be moving forward while the heel lifts, leading to an earlier detection. Some of the differences in this study compared to previous validations of CoP GED in YH (Roerdink et al., 2008) could, thus, be explained by differences between weight shift and actual movement.

Differences observed in detection of gait events between CoP and kinematics could affect calculations of gait phase durations. However, early detection of FC would lead to a longer stance phase which in-turn would be offset by a shorter swing phase (figure 3). This has been observed both in this study of SS and in previous validations of CoP GED in YH (Roerdink et al., 2008). Overall, phase duration calculations derived from GED agree within 100ms which is acceptable for most applications.

Given the limitations of kinematic GED methods noted, CoP GED may be a more appropriate way of detecting gait events in SS. However, SS in this study all had butterfly shaped cyclograms. Therefore, CoP GED algorithms might not work for SS with more severely affected gait (Wong et al., 2004). Future work of CoP GED for more severely affected gait could be validated by using a (fore aft) split-belt treadmill. The cyclogram could be computed by combining signals from the two force plates with gait detection on the basis of the magnitude of the ground reaction under each foot separately.

Conclusion

This study showed that CoP based GED agreed within 100ms with kinematic algorithms suggested for use with SS walking on a treadmill. The differences in GED methods reflect the differences between movement (kinematics) vs weight transfer (kinetics) and suggest CoP GED may be more appropriate for gait analyses of SS than kinematic methods; even when walking and varying steps.

Conflict of interest statement

We confirm that there is no conflict of interest with the current submission and a full review and understanding of copyright guidelines has been completed.

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